Methods for Clinical Evaluation of Noise Reduction Techniques at Computed Tomography

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PURPOSE

1.) To provide an overview of existing noise reduction strategies for low dose computed tomography (CT), including filtered back projection, image-space denoising, projection-space denoising, and iterative reconstruction.
2.) To review quantitative and qualitative tools and metrics for evaluating noise reduction methods.
3.) To highlight the strengths and limitations of individual noise reduction techniques using radiologist feedback and clinical examples.

OVERVIEW

- For a given diagnostic task and patient size, dose reduction is primarily limited by the image noise level and object detectability
- Data processing and image reconstruction methods may decrease image noise, improving image quality or allowing decrease in radiation dose.

In addition to conventional reconstruction filters (kernels) applied during CT data reconstruction[1], three primary techniques have been developed for controlling noise in the final images. These include:

- Image-space denoising[2,3]
- Projection-space denoising[4-6]
- Iterative reconstruction[7-10]

TECHNIQUES

IMAGE-SPACE DENOISING

Image-space denoising (ISD) is widely available on commercial scanners as well as from 3rd party vendors. Linear or non-linear filters are directly applied to previously reconstructed images to remove noise. Non-linear filters are currently used to improve image quality, which facilitates radiation dose reduction[2,3]. A fundamental trade-off exists between image noise and low-contrast lesion detectability, which is the primary limit to clinical utility.

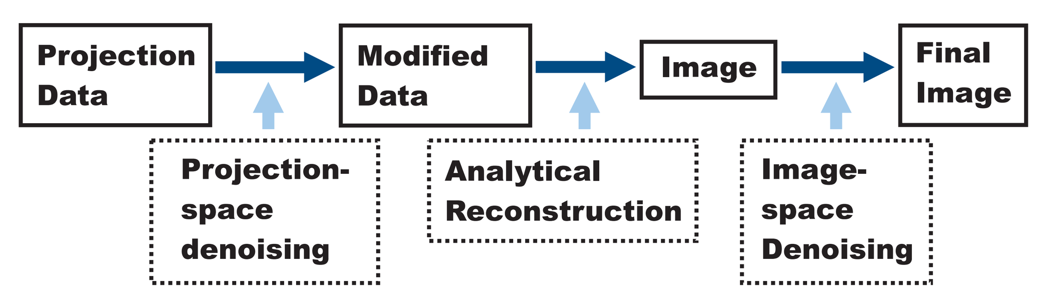

Figure 1. Projection- and image-space denoising applied during an analytical reconstruction.

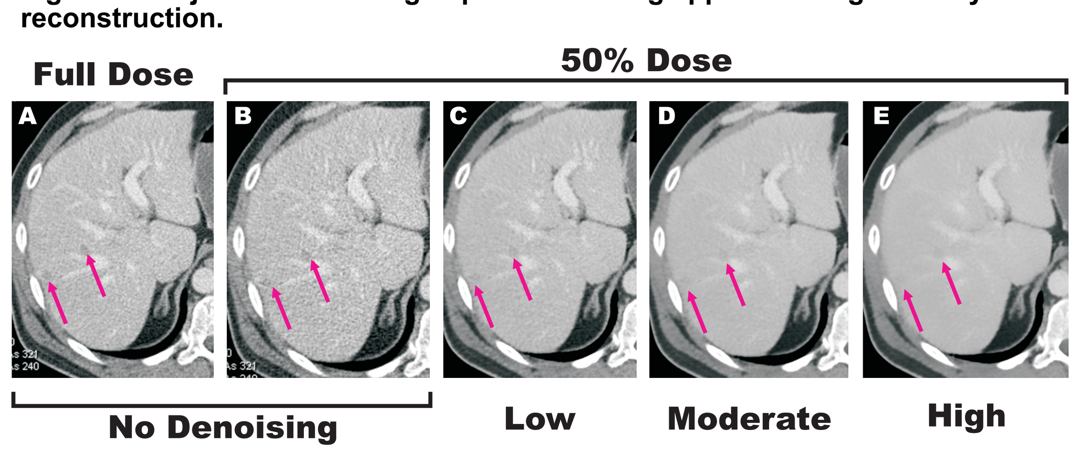

Figure 2. Two probable carcinoid liver metastases (arrows) at routine dose (A) and 50% dose (B). Three right images were reconstructed with different levels of denoising (C,D,E). Note that lesions become less conspicuous at high levels of denoising (D and E) and the images appear "cartoony."

PROJECTION-SPACE DENOISING

New techniques that remove noise from projection domain data prior to image reconstruction are termed projection space denoising (PSD)[4-6,11]. These methods account for photon statistics in CT data and smooth the data by optimizing a likelihood function and using a statistical noise model[4,5].

One recent approach for PSD is based on bilateral filtering, which smoothes projection space data using a weighted average, with weights based on spatial proximity and intensity of neighboring pixels, preserving important edge information[6]. Recent studies have shown the ability to reduce dose for abdominal and hepatic applications by imaging with lower kV CT and denoising the resulting data with projection-based denoising techniques to preserve diagnostic image quality and lesion conspicuity[12,13].

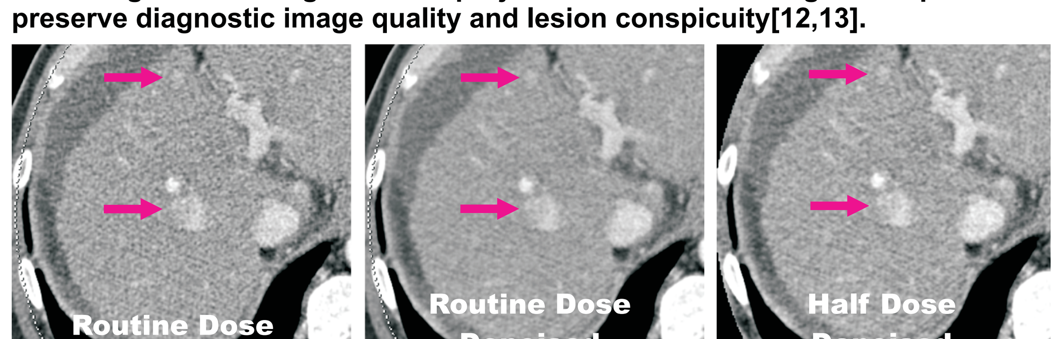

Figure 3. CT obtained during portal phase demonstrating two hepatocellular carcinomas (arrows). The routine dose image is the sharpest. The denoised routine dose image has higher CNR of the lesions due to noise reduction, but there is loss of image sharpness along the IVC border. Half-dose 80 kV image with denoising shows preserved HCC conspicuity with slight loss of image sharpness. (Reproduced with permission from Ehman et al. AJR Am J Roentgenol. 2012; 198:405[11].)

Although less pronounced than with image-space denoising, a drawback to projection-space denoising is potential loss of spatial resolution.

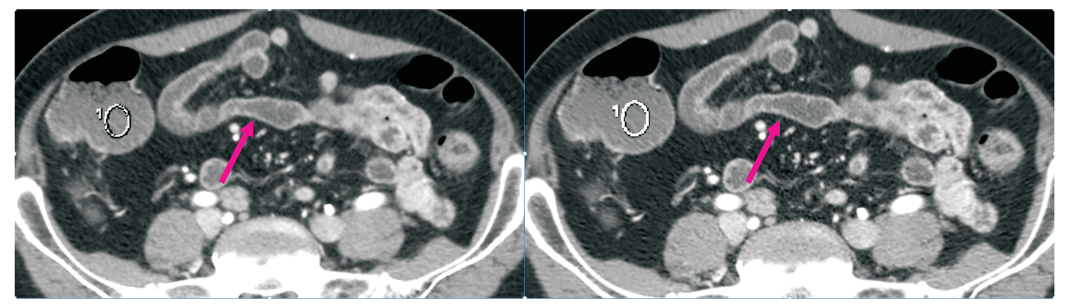

Figure 4. Images from low-dose CT enterography using a denoising method in projection space (left image) and Noise Map imaged-based denoising method (right image) at the same dose. Note blurring of the fat-wall interface of an ileal bowel loop (arrow) with the projection-based method.

ITERATIVE RECONSTRUCTION

Traditionally used in positron emission tomography (PET) and single photon emission computed tomography (SPECT)[14-16], iterative reconstruction (IR) has recently been applied to X-ray CT[7,8,10].

Compared with filtered back projection (FBP) techniques, IR has advantages in more accurately modeling the system geometry and noise, incorporating physical effects such as beam spectrum, beam hardening effect, scatter and incomplete data sampling[10,17]. Therefore, IR algorithms are better able to reduce output image noise while improving spatial resolution[8,10] and reducing artifacts[9,10,18].

Iterative reconstruction has significant potential to decrease radiation dose because of these advantages, but potential for dose reduction while preserving diagnostic accuracy has yet to be defined for different diagnostic tasks[19,20,23,24].

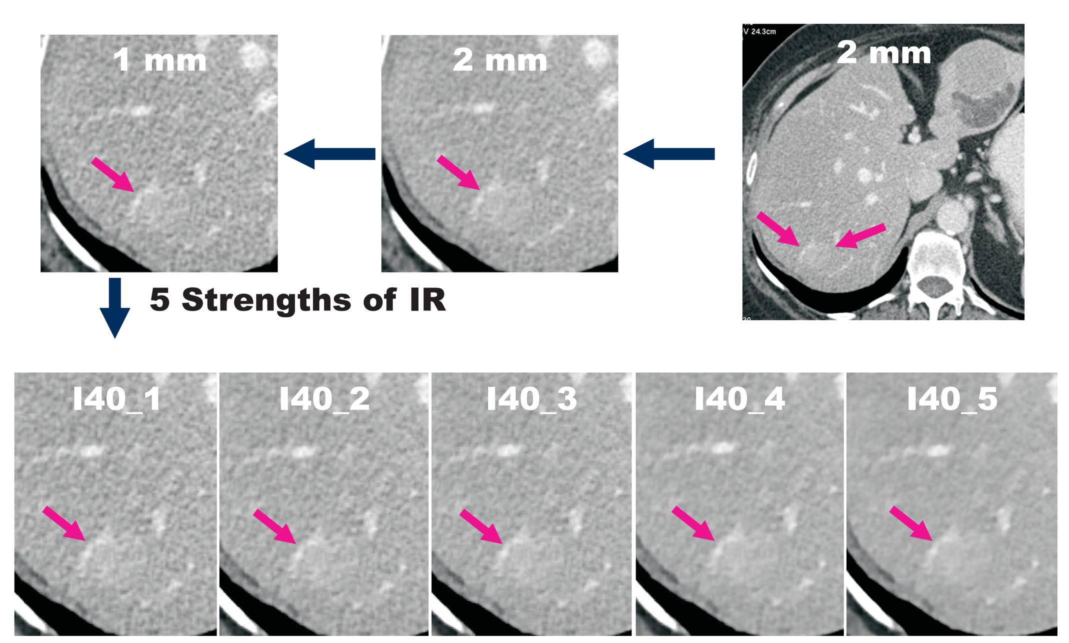

Figure 5. Subtle carcinoid metastasis (arrows) on routine dose 2 mm image (top right). A 1-mm slice is used to emulate noisier low-dose data and demonstrate the effect of SAFIRE (sinogram-affirmed iterative reconstruction; Siemens Healthcare). Lesion conspicuity and image sharpness is preserved, but some "cartooniness" and loss of normal noise texture occurs at the highest setting. I40_3 reduces noise to the level in the original image.

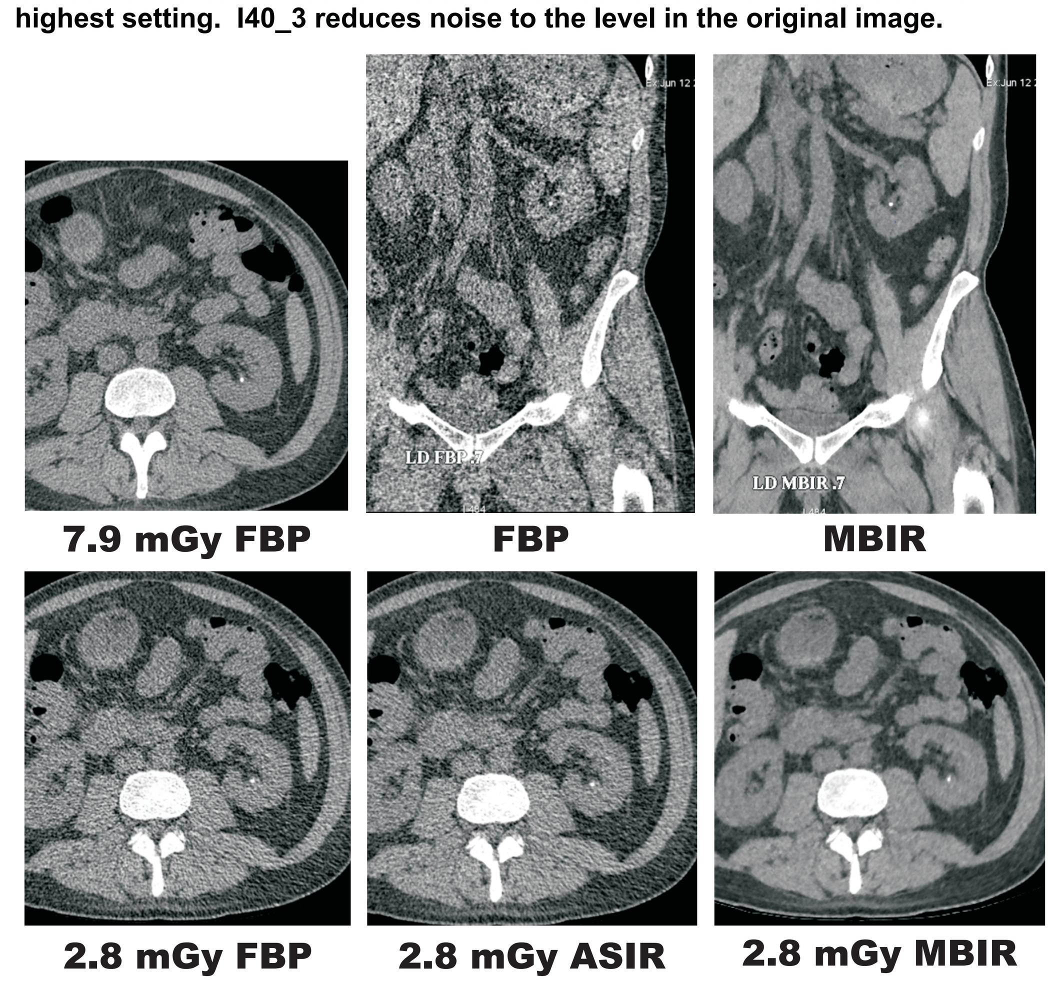

Figure 6. Left renal caliceal tip stone seen using 2-mm standard dose (7.9 mGy, upper left), and with 2-mm lower dose (2.8 mGy, lower row). Note improved noise reduction and improved stone conspicuity with MBIR compared to FBP and ASIR. Top row shows 0.7 mm coronal image with more dramatic improvement. Image quality improvement with IR and other denoising techniques is generally best using thinner images.

SUMMARY OF TECHNIQUES

Descriptions	Example methods	Advantages	Disadvantages
Traditional FBP-based methods	Weighted 3D FBP, AMBP	Fast; Directly available on scanners	Treats every ray the same, sub-optimal dose efficiency
Image-space denoising	SafeCT, SharpView, Noisemap NLM	Fast; Only needs reconstructed images, cross scanner platform	Does not take into account system physics; No capability to reduce artifacts
Projection-space denoising	Adaptive filtering or iterative denoising in projection data. *ASIR, ADIR, iDose might belong to this category, but no technique detail has been published.	Fast; May incorporate complex system physics models	Potential to lose spatial resolution if not designed well
Full iterative reconstruction	MBIR	Incorporates system model (both photon statistics and detailed geometry); has the potential to reduce both noise and artifacts	Slow; May change the noise texture of CT images
Hybrid iterative reconstruction	SAFIRE	Noise reduction in image-space (for speed), artifact reduction in projection-space (for image quality)	May change the noise texture of CT images

IMAGE QUALITY METRICS

- CT Number & Noise level – Accuracy and uniformity, assessed by ACR phantom
- High Contrast Spatial Resolution (Figure 7)
  - In-plane: Bar patterns – ACR phantom, MTF – Wire phantom
  - Cross-plane: Slice thickness patterns – ACR phantom
  - Slice sensitivity profile (SSP) – thin foil phantom
- Low Contrast Spatial Resolution (Figure 8) – low contrast ACR phantom
- Noise Power Spectrum (NPS) (Figure 9) – uniform phantom (30 cm water phantom); measures spatial frequency component of the noise.

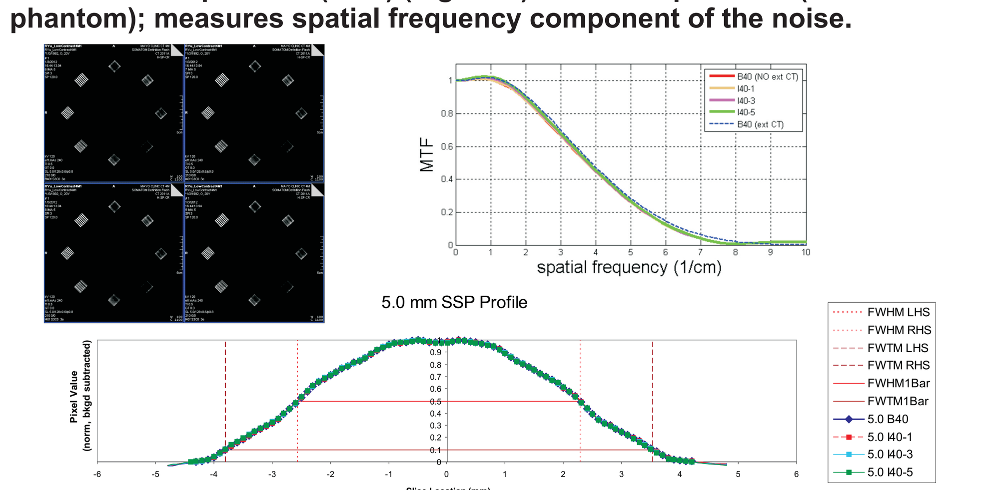

Figure 7. Qualitative in-plane high contrast spatial resolution using the bar pattern in the ACR phantom (upper left). A quantitative descriptor of in-plane spatial resolution is the MTF (upper right). Slice sensitivity profile (bottom) is used to measure spatial resolution in the z-direction. Note preserved high contrast spatial resolution with SAFIRE despite reduction in image noise.

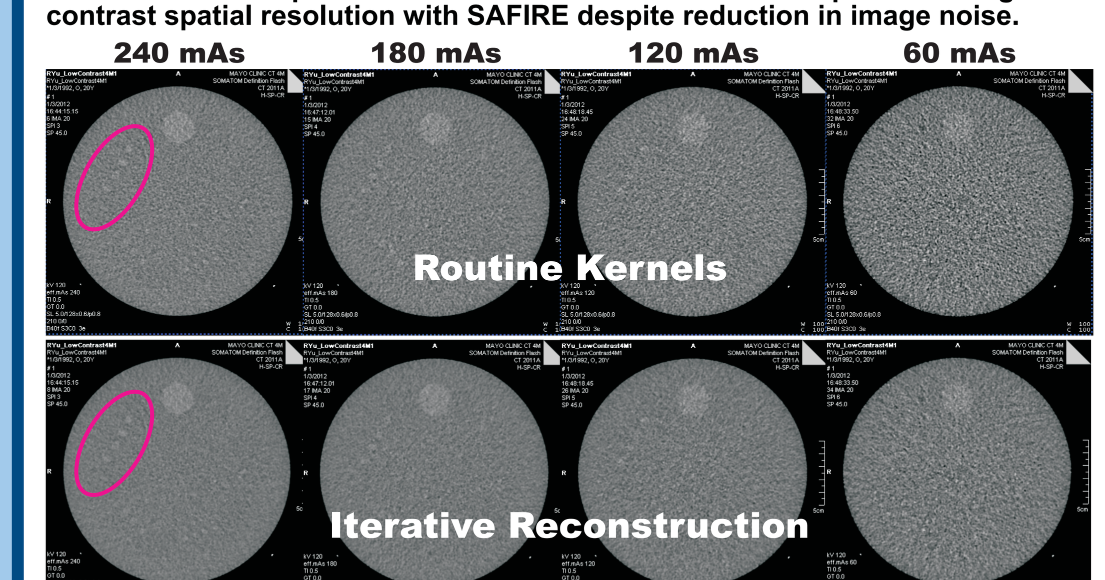

Figure 8. ACR low contrast phantom scanned at multiple dose levels shows that iterative reconstruction lowers noise level at all tube currents, but that small low contrast rods (circled on 240 mAs images) are difficult to detect at 120 and 60 mAs.

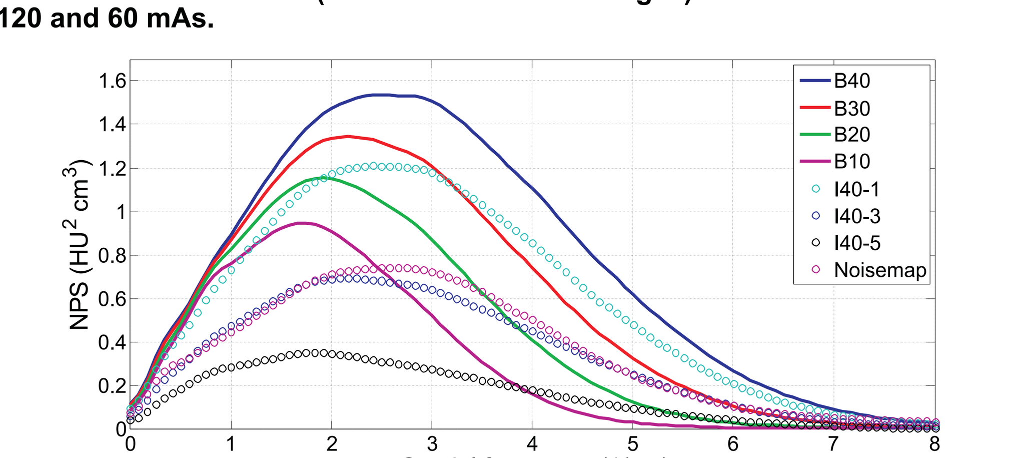

Figure 9. Noise power spectrum (NPS) is a more complete descriptor of noise properties than noise level. With IR noise reduction, the peak frequency of image noise shifts to the left as the strength is increased and noise decreased, which may correspond to a small sacrifice in low-contrast spatial resolution change noise texture.

Using these quality metrics, the trade-off between image noise and spatial resolution may be quantified for each reconstruction technique.

NON-LINEARITY OF PSD AND IR

- Due to non-linearity of iterative reconstruction and most denoising methods, spatial resolution is contrast- and noise- dependent
- Iterative reconstruction and non-linear denoising methods can substantially reduce the noise without sacrificing high-contrast spatial resolution (in terms of MTF and SSP), however, there may be loss of low-contrast spatial resolution

If dose reduction is not carefully controlled, imaging findings may be lost from the reconstructed images[19].

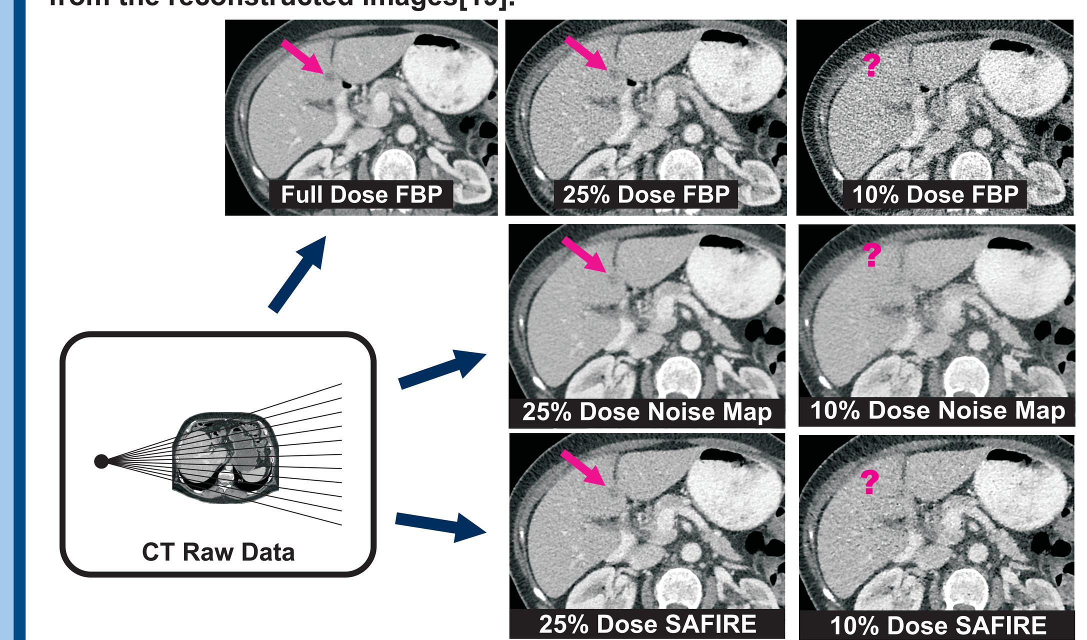

Figure 10. Images of a colon cancer metastasis (arrows) in the medial segment of the liver at routine dose (CTDIvol = 10.4 mGy), and 25% and 10% dose using routine reconstruction kernel (top row). Compared with Noise Map and SAFIRE reconstructed images at 25% and 10% dose. Note that the lesion is visible using all methods down to the 25% dose level, but is not visible using any reconstruction method at the 10% dose level.

CLINICAL EXAMPLES

As shown by phantoms, IR and non-linear denoising may maintain high-contrast resolution while simultaneously reducing image noise substantially. There is the potential for significant dose reduction for diagnostic tasks involving detection of high-contrast objects such as CT colonography, CT for renal calculus, and CT enterography for Crohn's disease.

IR and non-linear noise reduction methods may sacrifice low-contrast resolution while reducing image noise. This may correspond with more modest dose reduction for diagnostic tasks involving detection and characterization of low-contrast lesions such as those encountered in liver/pancreas or neuro- CT.

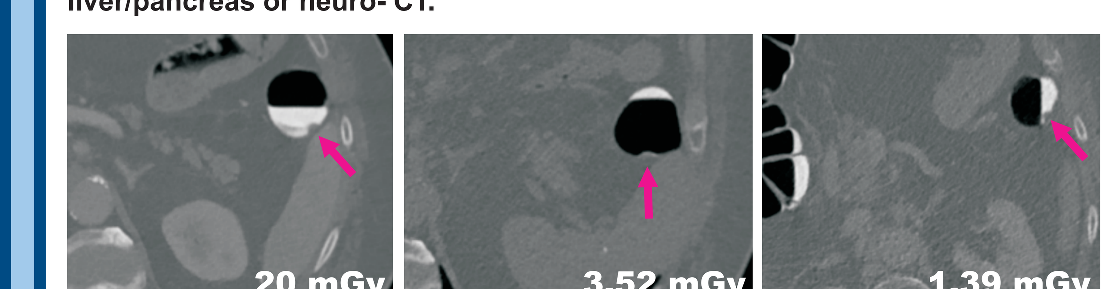

Figure 11. Images from a contrast-enhanced CT colonography study. Higher dose image (left) demonstrates a 0.7 cm polyp (a tubular adenoma) in the descending colon. The polyp is also seen on lower dose prone and decubitus images with SAFIRE despite > 80% and > 93% dose reduction. This is possible because colonography is a high contrast diagnostic task (i.e., differentiating polyp from air or contrast).

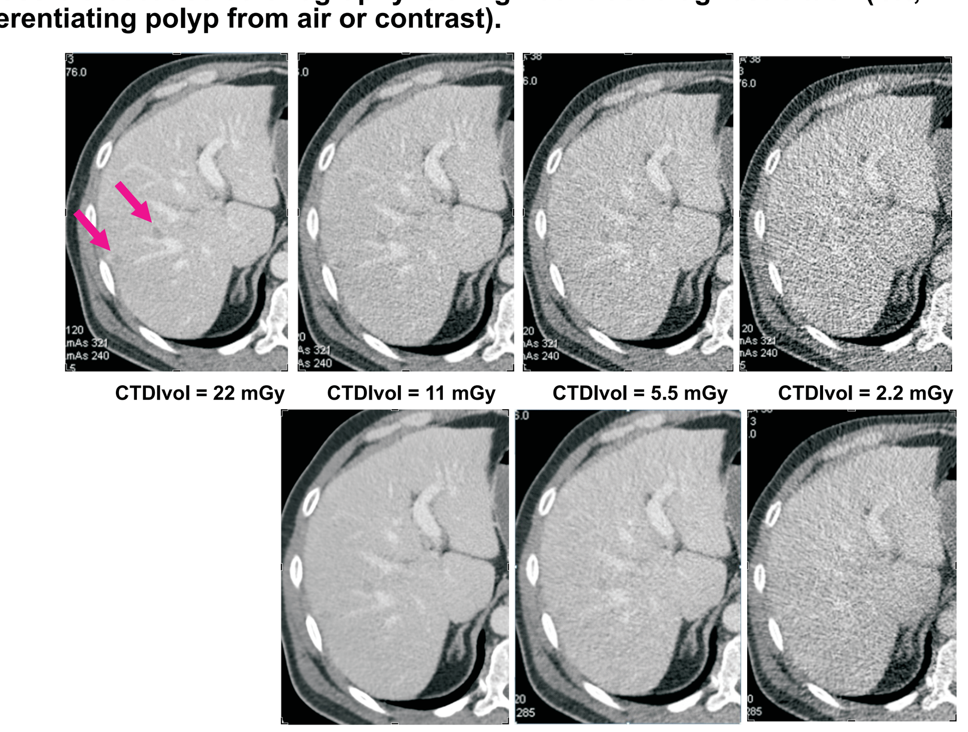

Figure 12. Same patient as Figure 3 with carcinoid liver metastases. Top row shows images reconstructed with FBP and routine dose, 50% dose, 25% dose and 10%. Bottom row represents image-based Noise Map denoising at corresponding dose levels. Note that neither FBP or denoised images display both metastases below the 50% dose level.

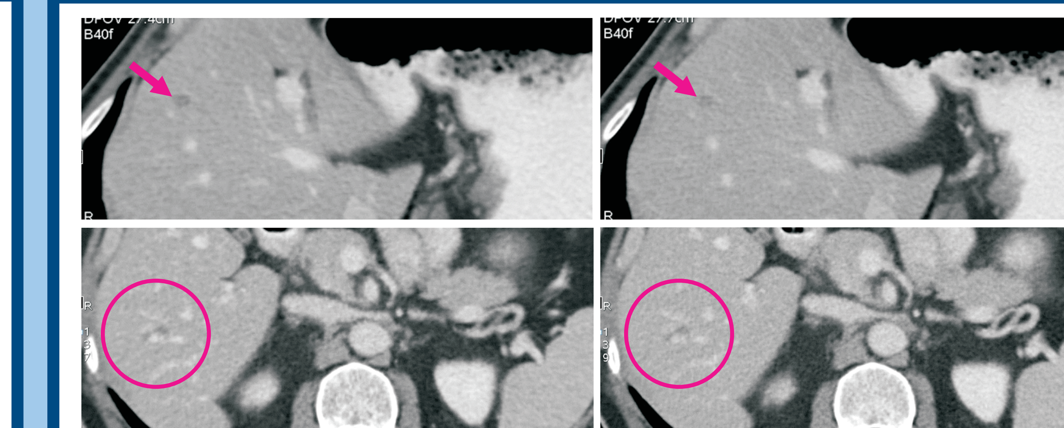

Figure 13. Comparison of Mayo Noise Map (left) and a commercial system (right). Note improved conspicuity to small metastasis (arrows) and hemangioma (circles) using Noise Map despite similar reductions in noise, but slightly decreased image sharpness and more coarse noise texture with the Noise Map method.

Because of the non-linear impact of denoising techniques on signal characteristics and spatial resolution, the effect of denoising on image texture and lesion conspicuity can differ between denoising methods and strengths. While these idiosyncratic effects can be troublesome to radiologists at first (like viewing images from another CT vendor or reconstructon kernel), they may or may not affect observer performance; familiarity with the techniques may lead to greater degrees of comfort.

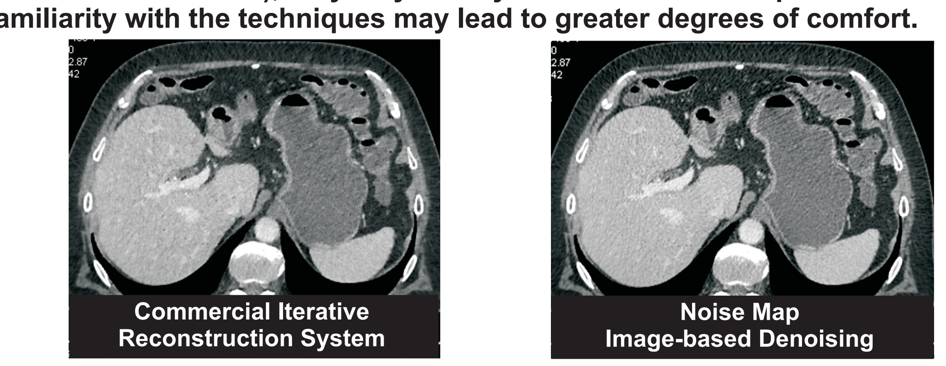

Figure 14. Figures from low-dose CT enterography in a large patient (CTDIvol ~ 4 mGy) show "pixelated" appearance to noise texture using a commercial iterative reconstruction system compared to an image-based denoising method. Effects of denoising on noise texture can be unpredictable and depend on dose level, strength of denoising, denoising method, and signal characteristics of the underlying organ.

CONLUSION

Multiple institutions have shown that careful integration of noise reduction strategies in to clinical practice can have a substantial effect on lowering radiation dose[13,19,22,24-26]. In order to preserve the diagnostic benefit of CT (and ensure accuracy for detection and diagnosis), careful optimization is needed when adopting these methods. Specific attention must be paid to the diagnostic task required for each patient. Potential dose reduction using noise reduction methods is dependent on the lesion-to-background contrast and the anatomy of interest. High contrast tasks such as CT colonography, renal stone detection, and CT enterography will permit large reductions in radiation dose. Low contrast tasks such as liver metastasis detection will permit smaller reductions in radiation dose. Methods to quantitate contrast-dependent spatial resolution, observer performance studies for a variety of diagnostic tasks (and tools to facilitate their rapid completion), and methods to predict the lowest dose to achieve adequate performance for specific CT systems are needed. The full impact of noise reduction techniques on radiation dose and radiologist performance is in the early phases of realization, with great potential to benefit patients by decreasing the radiation dose received while undergoing CT studies.

Summary

Various noise reduction methods have a great potential for reducing CT dose and improving image quality in clinical practice. Each noise reduction strategy has inherent strengths that should be employed and potential weaknesses that should be understood. While dose reduction may be aggressive for high-contrast tasks, careful optimization is needed for low-contrast tasks to prevent loss of lesion detectability.

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